Evaluation of PHBHHx and PHBV/PLA fibers used as medical sutures

Yu He · Zhiwei Hu · Mengda Ren · Changkun Ding · Peng Chen · Qun Gu · Qiong Wu

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Abstract Two types of fibers were prepared by using bio-based materials: a mono-filament made from poly(3-hydroxybutyrate-*co*-3-hydroxyhexanoate) (PHBHHx) and a multi-filament made from poly(3-hydroxybutyrate-*co*-3-hydroxyvalerate) (PHBV) and polylactic acid (PLA) blend. The two fibers were evaluated for mechanical properties, biocompatibility and degradability for the potential application as medical sutures. The PHBHHx fiber showed remarkable biocompatibility by H.E. Stainning, with very little impact to the surrounding tissues. The degradation of the fiber was observed by SEM after implantation for 36 weeks, and the major degradation product was detected after 96 weeks. Consistently, the PHBHHx fiber maintained more than half of the mechanical properties after 96 weeks. The other fiber was prepared by twisting PHBV/PLA blend strands to a bunch, and showed high biocompatibility and relatively high degradability. The bunched structure loosed after 36 weeks of implantation. These low-cost and easily prepared fibers have great potential in medical applications, since they could avoid the formation of fibrous capsule, reduce the size of scar, and degrade into non-toxic and even beneficial products.

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1 Introduction

Medical sutures are sterile fibers used for ligation, hemostasis and tissue fixation, and play a significant role in early wound-healing stage. They are simple but important articles used in medial operation with a fast-growing market [1, 2]. Depending on whether the body can naturally degrade and absorb the suture, medical sutures can be divided into absorbable and non-absorbable sutures. Absorbable sutures, previously defined as sutures which can lose most of the tensile strength within 60 days postimplantation [3], are able to degrade in vivo to form non-toxic breakdown products, thus reducing the risk of infection by avoiding a secondary operation.

An ideal absorbable medical suture should have the following characteristics [4–6]: smooth surface to avoid scratch on the surrounding tissues; ease to knot and handle; high biocompatibility, i.e. non-toxic with minimal tissue reactions and minimal influences on the surrounding tissues; resistance against bacterial growth; high tensile strength to ensure the closure of wounds; ease to sterilize; absorbed after serving its function, i.e., losing strength after the wound fully recovers. Therefore, specific absorbable sutures should be chosen according to the requirements of different wounds. In most cases, the rapid-healing tissues require fast-degrading sutures, but the long lasting sutures

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are apparently favorable for slowly healing tissues such as fascia, bone and tendons [1, 3, 7, 8].

Based on their sources, absorbable sutures can be divided into natural sutures and synthetic sutures. Catgut is the most commonly used natural absorbable suture due to its diverse sources and low cost, but it also has many shortcomings. First, catgut inclines to cause serious tissue reactions [9–12]; second, its mechanical properties such as low flexibility cannot fulfill the requirements of different operations; also, it would lose all the tensile strength within a few days [5]. Other natural fibers such as collagen fiber, chitin fiber and chitosan fiber were also invented, but they usually need specific processing technology. Furthermore, chitin and chitosan have complex degradation mechanisms and may cause severe immune reactions to some populations [13, 14].

Synthetic absorbable sutures are fibers produced from strands of synthetic polymers, and they are usually easier to process than natural fibers. As catgut can elicit intense tissue reactions, polyglycolide (PGA) suture with higher biocompatibility was developed in 1968, which was the first synthetic absorbable suture [3, 5, 15, 16]. PGA suture also has better mechanical properties and handling characteristics than catgut, but it lacks flexibility and its rough surface may scratch the surrounding tissues. It may also cause slight tissue reactions [17, 18].

Another synthetic absorbable suture poly(lactide-*co*-glycolide) (PLGA) was developed in 1974 by random ring opening polymerization of two monomers, the cyclic dimers of glycolic acid (GA) and lactic acid (LA). Different degradation rates of PLGA can be obtained depending on the LA/GA ratio [19]. Two other types of synthetic absorbable sutures viz. Polydioxanone (PDS) and polylactide (PLA) were developed in 1981, which overcame one major disadvantage of previous synthetic sutures, the handling problem. The smooth surface and soft texture allowed easy bending and knotting, which made the handling process simple and safe. Also, with improved biocompatibility, PDS suture could be used in vascular and nervous tissues. However, the biocompatibility and mechanical properties of these sutures still need further improvement to meet the various requirements of different wounds. Therefore, other sutures made of biopolymers such as polyhydroxyalkanoates (PHAs) have drawn great attention these years for they can offer properties not available in the existing synthetic absorbable polymers [20].

PHAs are a family of intracellular polyesters synthesized by a variety of bacteria, which usually serve as the carbon source and energy storage material [21, 22]. The general structure of PHAs is shown in Fig. 1. PHAs materials are biodegradable and highly biocompatible, and the first-generation product polyhydroxybutyrate (PHB)

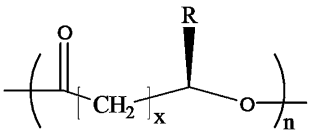


Fig. 1 The structure of Polyhydroxyalkanoates (PHAs) PHAs are biopolymers synthesized by microorganisms. $x = 1, 2, 3\ldots$, usually $x = 1$, i.e. poly 3-hydroxyalkanoates. Different R groups indicate different monomers of PHAs: for 3-HB, 3-HV and 3-HHx, R=CH_3, C_2H_5 and C_3H_7, respectively

have already been used as sutures [11, 23]. In comparison, the second-generation poly(3-hydroxybutyrate-*co*-3-hydroxyvalerate) (PHBV) has higher flexibility, thermal stability and processibility, making it promising as a medical suture. When PHB and PHBV were implanted to animals intramuscularly, the tissue reaction was similar to the reaction to silk and less severe than the reaction to catgut [11, 23]. Poly(3-hydroxybutyrate-*co*-3-hydroxyhexanoate) (PHBHHx) is considered as the third-generation PHA, which consists mainly of 3-HB (3-hydroxybutyrate) and partially of 3-HHx (3-hydroxyhexanoate). The large-scale production of PHBHHx has been accomplished after years of trials [24]. This polymer shows higher break strength and flexibility than PHB and PHBV [25], greatly higher biocompatibility and hemocompatibility [26, 27], and high affinity to a variety of cells [28–30], which makes it even more advantageous for medical sutures application.

However, in spite of their biodegradability, high biocompatibility and their unique properties, PHBV and PHBHHx could be further modified for better applications as medical sutures. Previous studies on PHBV and PHBHHx films indicated that the compositions of the monomers would greatly affect the mechanical properties of the polymer [25, 31, 32]. For example, the elastic modulus and toughness were enhanced and the crystallization rate was reduced as the 3-HHx content in PHBHHx was increased. Therefore, the compositions should be appropriately designed for the preparation of sutures. In this study, the compositions for preparation of PHBHHx and PHBV fibers (see. Sect 2) could balance the mechanical properties and crystallization rate [33, 34]. Twisted multi-filaments with ultra-thin strands usually had better properties and higher handle ability. With PLA (72 wt% of the total fiber) added into PHBV, the resulting PHBV/PLA blend strands have better handle ability [35, 36].

In this research, we used PHBHHx and PHBV/PLA fibers as medical sutures and implanted them in rats for as long as 96 weeks. Their qualities were evaluated during the process, including mechanical properties, biocompatibility, degradation, and tensile strength maintenance. Those results suggest that these two biocompatible materials could be used as medical sutures for further studies.

2 Methods and materials

2.1 Preparation of fibers and measurement of mechanical properties

The materials used to spin strands in this study are PHBHHx (12 mol % 3-HHx, 88 mol % 3-HB) and PHBV/PLA (28 wt% PHBV with 1.2 mol % 3-HV/98.8 mol % 3-HB, blended with 72 wt% PLA). The PHBHHx fiber was prepared according to the patents [37, 38] and Ref. [39], and the PHBV/PLA strands were prepared by Ningbo Tian'an Institute of Materials [36].

The diameters of the strands were measured by using scanning electron microscope (SEM), and other mechanical properties were measured with an electronic pull test machine (GOTECH, AI 7000S). The drawing speed was 50 mm/s and the interval of fixtures was 15 cm.

2.2 Implantation

Three-month-old male Sprague–Dawley (SD) rats (180–200 g) were used and randomly divided into 16 groups each having 3 rats, with 8 groups for implantation of PHBHHx and 8 groups for PHBV/PLA. The rats were injected with pentobarbital into the celiacs as anesthetic. The hair on the back near the tail was scraped with blades to expose the skin, and then the skin was incised longitudinally from the middle. After that, a syringe needle inserted with sutures was used to impale the tissues, and the needle was pulled off the tissues quickly to let the fibers remain in the tissues. Medical catguts were used as control. The PHA sutures (left side) and catguts (right side) were implanted symmetrically in the subcutaneous tissues and muscle tissues (Fig. 2).

2.3 Histological analysis

After 1, 3, 6, 12, 24, 36, 48, 96 weeks of implantation, 3 rats were randomly chosen as a group and injected with overdose pentobarbital. The chest was cut open to puncture the left atrium, and 40 ml of formalin (10 % Formaldehyde, PBS buffer) was injected into the right ventricle. After formalin circulated the whole body and reached the left atrium, the hair on the back near the implantation site was scraped and the strands along with the surrounding tissues were taken out. Then 2/3 of the samples were fixed in 10 % neutral buffered formalin for 24 h and embedded in paraffin for hematoxylin-eosin (H.E.) Staining, while the other samples were used in other experiments such as molecular weight measurement, retained tensile strength measurement and 3-HB assay. Each sample was cut at 3–5 positions and the resulting 5-μm thick sections were used for H.E. Staining. The paraffin embedding and H.E. Staining were performed in the Center of Biomedical Analysis, Tsinghua University.

2.4 Molecular weight measurement by gel permeation chromatography (GPC)

SHIMADZU Prominence GPC analyzed system (LC-20AT Solvent Delivery Uni, RID-10A Differential Refractive Index Detector, Column Oven CTO-20A, and Auto-sampler SIL-20A) was used to measure the molecular weight of the samples. The PHBHHx strands after implantation for 36 weeks were lyophilized and dissolved in chloroform, and after filtration through a 0.22-μm filter, the molecular weight was measured using GPC. The non-implanted PHBHHx strands were used as control.

2.5 Surface observation by SEM

Scanning electron microscope (FEI Quanta 200, Center of Biomedical Analysis, Tsinghua University) was used to observe the surface of the PHA fibers. All samples were sputter-coated with gold for 60 s before observation.

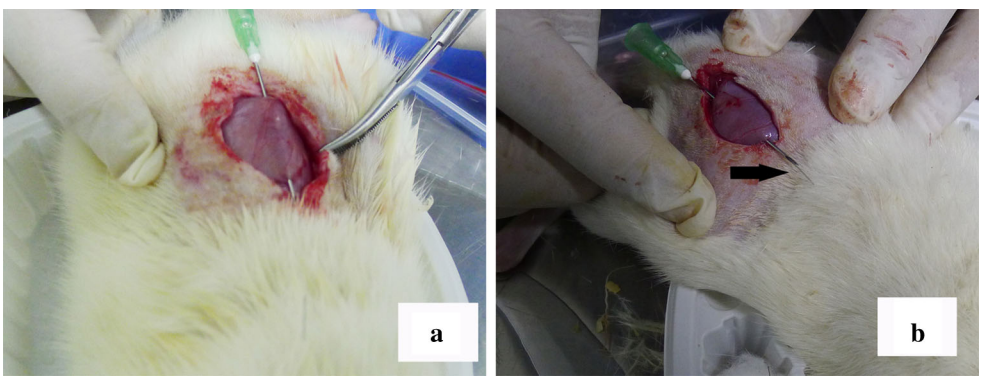


Fig. 2 Procedures for implantation of PHBHHx and PHBV/PLA fibers. **a** PHBHHx or PHBV/PLA fiber was implanted in the *left side*. **b** Catgut was implanted in the *right side*

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2.6 3-HB monomer assay

The muscle tissues where the PHBHHx fiber was implanted were analyzed to detect the concentration of 3-HB. Three pieces of a muscle tissue were used: one piece in the back which was in direct contact with the fiber, one piece in the back near the implantation site, and one piece in the leg. Two parallel samples were analyzed for each group.

The concentration of 3-HB was determined by using β-hydroxybutyrate (Ketone Body) Colorimetric Assay Kit (Cayman Chemical Company, America). The samples were prepared as follows: First, 400 mg of muscle tissue was obtained from each sample, and then minced into small pieces. Each tissue was immersed with 200 µl of an assay buffer, and then the minced tissues were homogenized with an OSE-Y10 electric tissue grinder (Tiangen, China) in 1.5-ml centrifuge tubes. Another 800 µl of the assay buffer was added, following the protocol of the kit.

3 Results

3.1 Preparation and mechanical properties of PHBHHx and PHBV/PLA fibers

Figures 3 and 4 showed the structures of the PHBHHx fiber which was a monofilament and the PHBV/PLA fiber which consisted of 20 twisted strands respectively. The diameters and other mechanical properties of the strands in the fibers were listed in Table 1. The very thin PHBV/PLA strand was relatively easier to crystallize, so it was stretched with the help of an automatic spinning machine. However, it was difficult to draw PHBHHx strand in a common way due to its high viscosity and low crystallization rate. Therefore, a semi-automatic machine was used for the preparation of PHBHHx fibers [39]. Even better mechanical properties of PHBHHx fibers could be expected if an automatic spinning method was developed. Both PHBHHx and PHBV/PLA fibers have regular structures with smooth surfaces and uniform diameters (Figs. 3a and 3b), and they were easy to stain according to our previous studies (data not shown), which allowed them to be used as medical sutures.

Fig. 4 Morphology of muscle tissues surrounding PHBV, PHBHHx ▶ and catgut fibers. The muscle tissues with fibers implanted were taken out at different time points. 5-µm thick sections were cut for H.E. Staining. The *left, middle* and *right columns* were tissues with PHBHHx, PHBV/PLA and catgut fibers implanted, respectively. *Six lines* represented different implantation periods (1, 3, 6, 12, 24, 48 weeks). Labelling *m* macrophage, *mf* muscle fibers. H. E. Staining was used for visualization. PHBHHx and PHBV/PLA fibers showed much less tissue reactions, indicating much higher biocompatibility than catgut. *Magnification* 10 × 40; *Scale bar* 100 µm

3.2 Implantation and biocompatibility

Both the PHBHHx and PHBV/PLA fibers were used as medical sutures for operations in muscular and subcutaneous tissues in rats, as described in see. Sect 2. Catgut was used as control. The operations went smoothly with neither

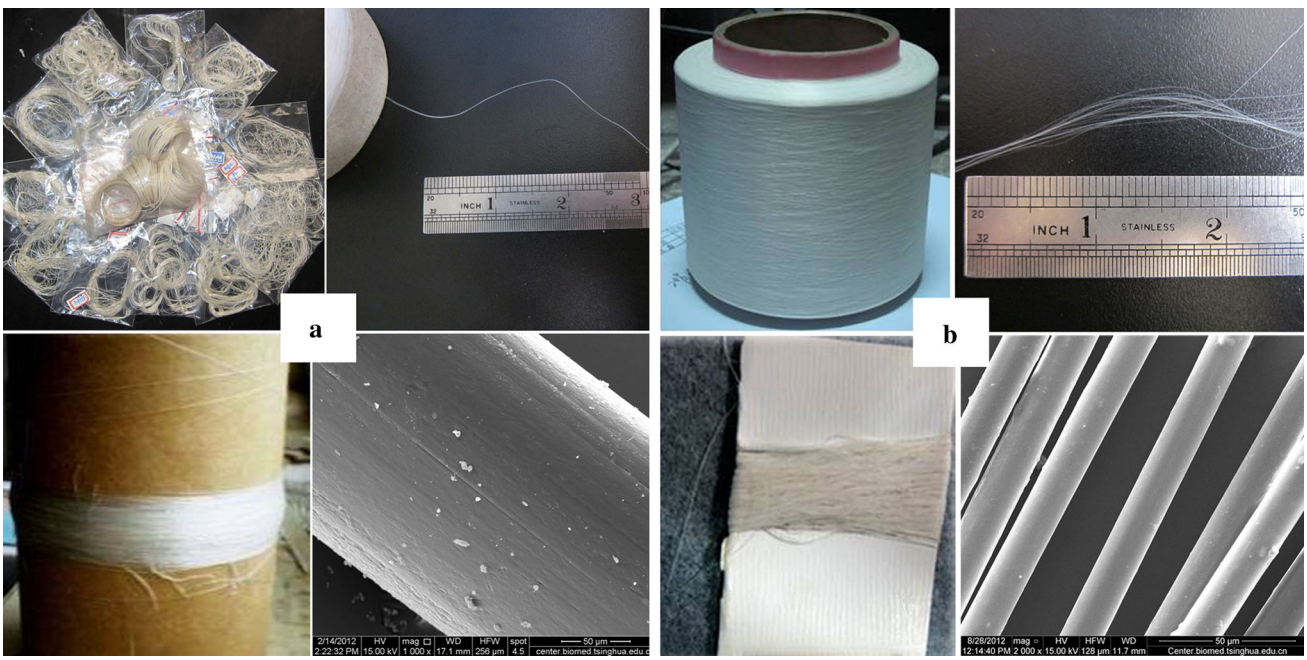


Fig. 3 Morphology of PHBHHx and PHBV/PLA fibers. Two sorts of PHA fibers were prepared for use as medical sutures. **a**, **b** show the morphologies of PHBHHx fiber and PHBV/PLA fiber respectively. The detailed morphology was captured by SEM at *lower right corner*

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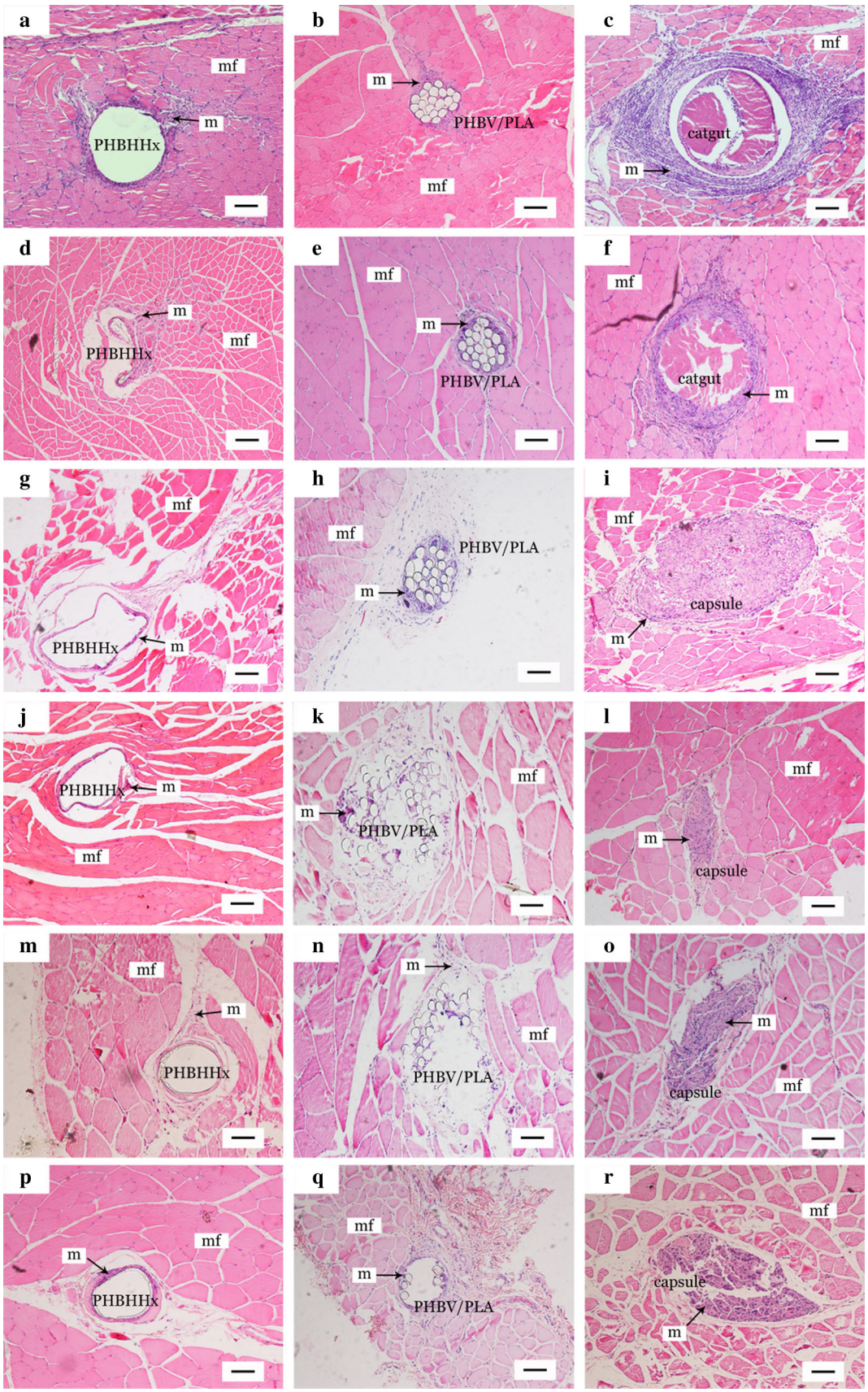


Table 1 The mechanical properties of PHBV/PLA and PHBHHx fibers

Fibers	Diameter (μm)	Maximum draw ratio	Tensile strength (N)	Elastic modulus (Mpa)
PHBV/PLA	14.5 ± 1.1	1.45 ± 0.07	0.13 ± 0.01	282 ± 42
PHBHHx	214 ± 17	1.59 ± 0.14	3.04 ± 0.64	289 ± 58

toxic symptom nor tumor at the implantation sites. Muscular and subcutaneous slices (5 μm) were taken out at different time points and stained for observation (Figs. 4 and 5).

After implantation for 1 week, apparent tissue reactions could be observed when catgut was implanted in muscular or subcutaneous tissues, with plenty of inflammatory cells (mostly lymphocytes and monocytes, and a few macrophages and giant cells) penetrating into the implantation area (Figs. 4c and 5c). The inflammatory cells (dark stained spots indicated by arrows) gathered around the implantation sites. The PHBHHx and PHBV/PLA fibers caused much less severe inflammatory reaction, indicating that biocompatibility was greatly improved (Fig. 4a). Catguts had already degraded at week 1, with inflammatory cells invading inside (Fig. 4c), while no significant degradation of PHBHHx or PHBV/PLA was observed. The PHBV/PLA fiber showed higher cell affinity than the PHBHHx fiber (Fig. 4a, b), which was consistent with a previous study [40].

Three weeks after implantation, the catgut was entirely digested into pieces and the implantation area was large (indicated by arrows in Figs. 4f and 5f), indicating severe inflammation. The PHBHHx and PHBV/PLA groups maintained their shapes, with slight tissue reaction (Figs. 4d, e, 5d, e). Similar phenomena were observed at 6 weeks after implantation, when the catgut fiber was completely degraded and the interspace was filled by collagen and fibroblasts to form a spindle-shaped fibrous capsule (Fig. 4i). On the contrary, the PHBHHx fiber merely degraded, and collagen and fibroblast gathered outside the implantation site. The inflammatory area in the PHBV/PLA group expanded slightly, suggesting that the fiber started to degrade. Also, the twisted strands started to loose (Fig. 4h). From week 12 to week 48, the spindle-shaped fibrous capsules entirely replaced the catguts, similar to the results at week 6 (Fig. 4l, o, r), and the PHBV/PLA fiber further loosed and totally lost its bundle structure (Fig. 4k, n, q). However, the diameter of each remaining strand almost unchanged, indicating that the degradation rate was relatively low, which is similar with PHBHHx implanted in muscular and subcutaneous tissues (Figs. 4j, m, p, 5j, k, n).

Fig. 5 Morphology of subcutaneous tissues surrounding PHBV, PHBHHx and catgut fibers. The subcutaneous tissues with fibers implanted were taken out at different time points. 5 μm thick sections were cut for H.E. Staining. The *left, middle and right columns* were tissues with PHBHHx, PHBV/PLA and catgut fibers implanted in, respectively. *Six lines* represented different implantation periods (1, 3, 6, 12, 24, 48 weeks). After 12 weeks, PHBV/PLA fibers loosed in subcutaneous tissues and were hard to be observed. So the results were showed solely in three photographs (PHBV/PLA implanted for 1, 3, 6 weeks). Labelling: *m* macrophage; *mf* muscle fibers; *fc* fibrous capsule. H. E. Staining was used for visualization. PHBHHx and PHBV/PLA fibers showed much less tissue reactions, indicating much higher biocompatibility than catgut. *Magnification* 10 × 40; *Scale bar* 100 μm ▶

3.3 Biodegradation of the PHBHHx fiber

Degradation of the PHBHHx fiber did occur, although not apparent from the tissue slices. After 36 weeks of implantation into muscle tissues, slight erosion appeared on the surface of the PHBHHx fiber (Fig. 6b). There were stripped defects on the edge and the degradation was heterogeneous. As a result, the diameter of the fiber was decreased from the original one (Fig. 6a, b).

To further investigate the degradation process, the molecular weight of the PHBHHx fiber was measured by using GPC. After 36 weeks of implantation, the number-average molecular weight (nAMW) was reduced slightly, but the weight-average molecular weight (wAMW) remained the same (Table 2). The reason might be that the low-molecular-weight components in PHBHHx degraded faster than the high-molecular-weight counterparts, so nAMW was reduced faster than wAMW.

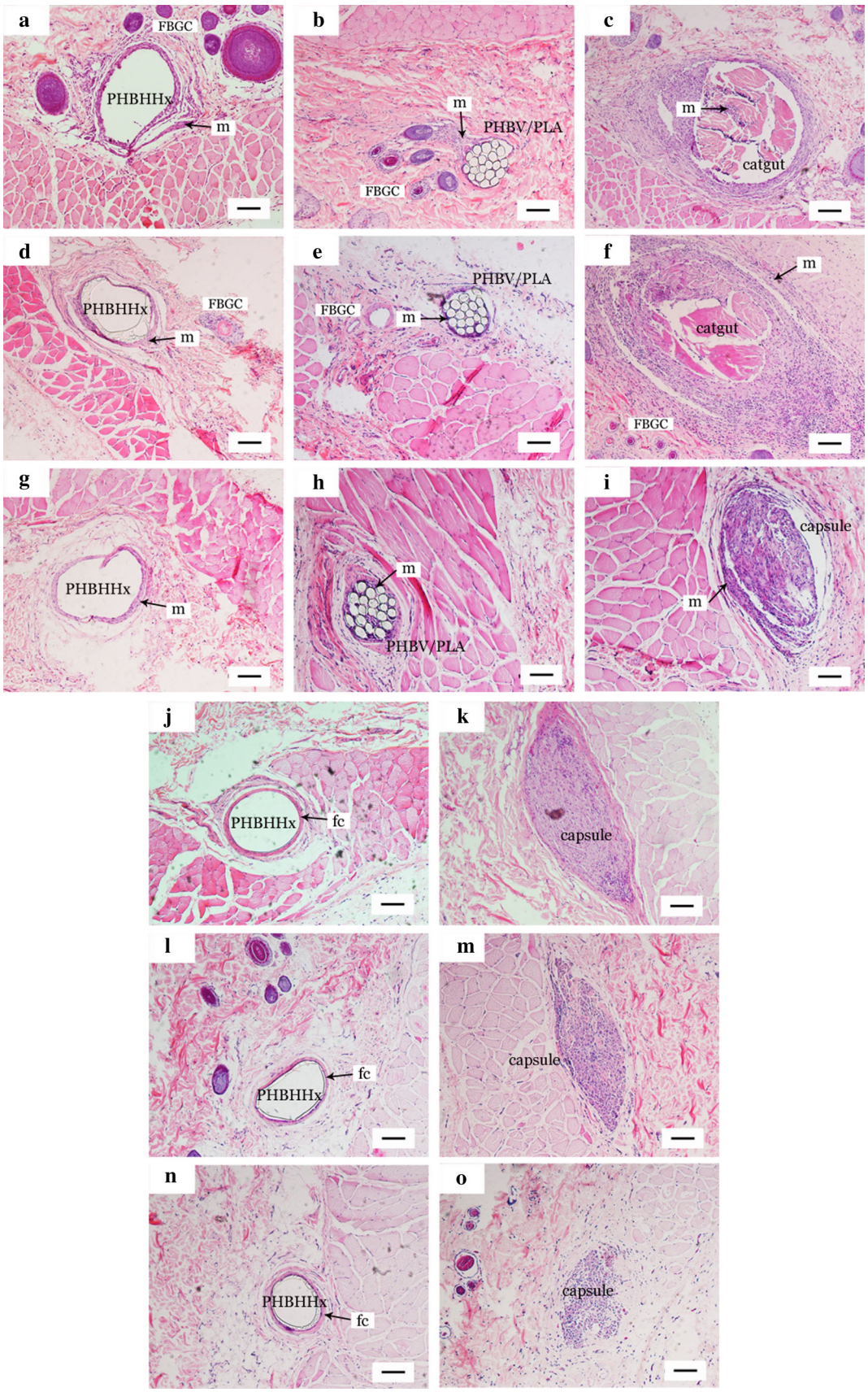
The concentration of 3-HB was detected after PHBHHx was implanted in muscles for 96 weeks. Samples were taken from tissues at different distances from the PHBHHx fiber (Fig. 7). The adjacent muscle samples contained higher 3-HB content than the distant ones, indicating that the degradation process resulted in a concentration gradient of the released 3-HB.

3.4 Tensile strength maintenance in muscle tissues

In selection of a suture, the properties of the suture must match the requirements of the tissue [41]. The mechanical properties of the PHBHHx suture, including the retained tensile strength, the maximum stretch ratio and the elastic modulus, were measured after 0, 24, 48, or 96 weeks of implantation (Fig. 8). The retained tensile strength and maximum stretch ratio were stable within the first 24 weeks, and then dropped after 48 weeks. After 96 weeks, the PHBHHx fiber still maintained more than half of its tensile strength, indicating a possible application for chronic wounds.

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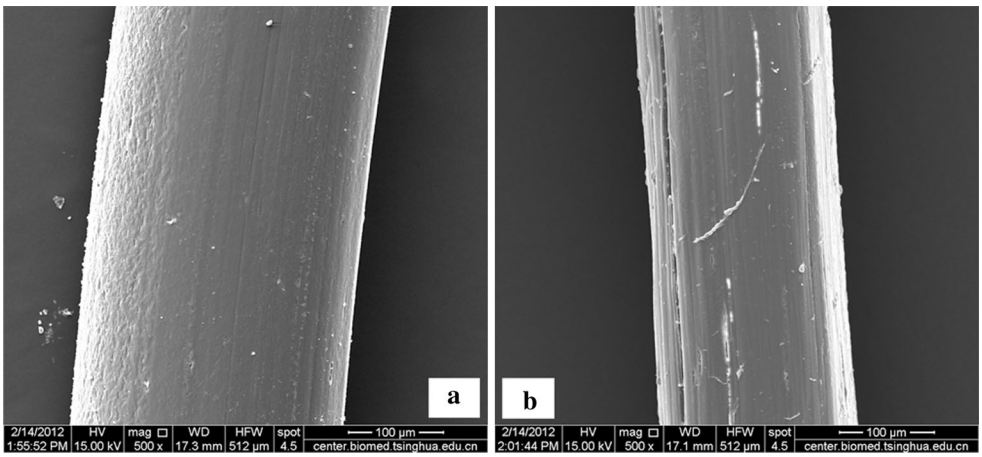


Fig. 6 PHBHHx strand degradation evaluated by SEM. After 36 weeks of implantation, the fibers were taken out and captured by SEM to observe the degradation. **a**, **b** show PHBHHx fibers unimplanted and implanted for 36 weeks. After implantation, PHBHHx fiber showed erosion on the surface, and its diameter was reduced, which showed the degradation of PHBHHx

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Table 2 Molecular weights of PHBHHx fibers with and without implantation measured by GPC

Fiber	Mn (kDa)	Mw (kDa)	Mw/Mn
PHBHHx fiber without implantation	184	283	1.54
PHBHHx fiber implanted for 36 weeks	177	283	1.60

Mn number-average molecular weight, *Mw* weight-average molecular weight, *n* heterogeneity

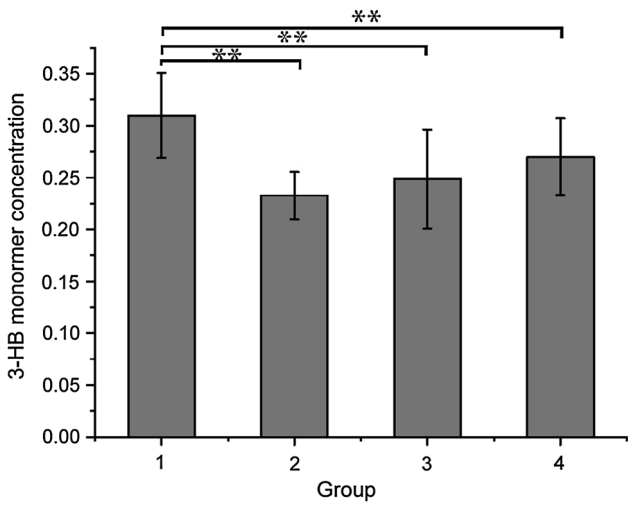


Fig. 7 3-HB concentration assay in different positions. The 3-HB (catabolite of PHBHHx) concentration was measured to evaluate the degradation. Group: 1 is for the tissue adhering to the PHBHHx fiber, 2, 3 are in the same muscle tissue as 1 but not adhering to PHBHHx fiber. 4 is for other muscle tissue with no PHBHHx fiber implanted in. $^{**}P < 0.01$

4 Discussion

One major problem of the existing medical sutures is the tissue reactions due to unsatisfied biocompatibility. PHA-based sutures, such as PHBHHx and PHBV used in this study, have superior biocompatibility. Their degradation products including 3-HB, 3HV and 3-HHx are totally non-toxic and even beneficial for the growth of vascular smooth muscle cells [42]. In this study, the PHBHHx and PHBV/PLA fibers were evaluated for application as medical sutures, and were both proven to have high biocompatibility.

According to our results, the PHBHHx fiber is a promising material for cosmetic surgical suture. Its high elasticity could accommodate to the deformation of the wounds; its high biocompatibility could avoid tissue reactions and reduce the irritation to the surrounding area. Also, the PHBHHx fiber maintained more than half of its tensile strength after implantation in body for a long time, and its high elasticity helps to keep the original shape and fix the wound. Therefore, this type of suture is particularly suitable for treatment of chronic wounds such as bone tissue repair or Achilles tendon repair. Applications in other tissues are also possible. Reportedly, PHBHHx showed high hemocompatibility [43], and also has great potential to be used in vascular closure. Besides, PHBHHx could accelerate the growth and differentiation of nerve stem cells [44], and PHBHHx-woven conduits were used to repair the nerve conduits [45]. These findings indicate that PHBHHx can be used as nerve repair material.

Although PHA-based sutures have many advantages including high biocompatibility and elasticity [46–48], they need further improvement, especially the handle ability. Also, due to the low crystallization rates of PHBV and PHBHHx, it was difficult to draw very long and thin fibers with uniform texture, which limited their applications. PLA is relatively more prone to crystallization and degradation than PHBV [49], and films and electro-spinning scaffolds made of PHBV/PLA and PHBHHx/PLA have been explored [50]. To overcome the drawbacks of PHA-based

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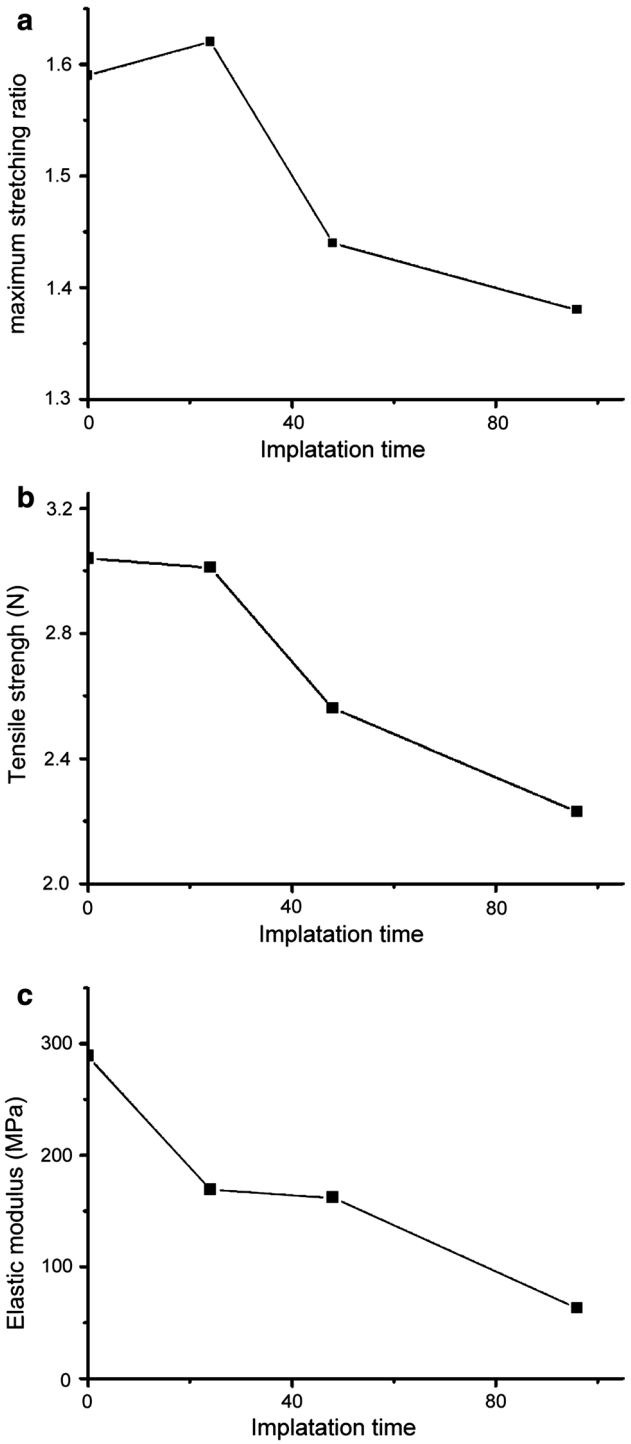


Fig. 8 The retained tensile strength of PHBHHx implanted for 24, 48, 96 weeks. The retained tensile strength of PHBHHx fibers was measured at different implanting time points. **a** change of maximum stretch ratio with time. **b** change of maximum tensile strength with time. **c** change of elastic modulus with time

sutures, PLA was blended into PHBV to accelerate crystallisation and enhance the handle ability. As a result, very thin strands with a diameter about 14 μm were produced, and by bundling 10 strands together, the fiber kept stable mechanical properties while maintaining a small overall diameter. Using this method, these materials can also be made into other medical items such as patches, artificial vascular grafts, or artificial nerve conduits. Thin fibers of the blends could be directly weaved into different shapes, avoiding any re-heating which would greatly damage the mechanical properties of the blends [51]. While maintaining high biocompatibility, the addition of PLA also greatly improved the handle ability of the suture, which resulted in smoother surface and easier knotting and thus would make operations simpler and less risky.

Certainly, there are some limitations for these fibers to be used as medical sutures. Both PHBHHx and PHBV/PLA fibers are hydrophobic, which would reduce cell affinity. It is generally considered that a higher hydrophilicity will provide a suture with higher cell affinity [52]. The hydrophilicity of these fibers can be improved by many methods, such as treatment with NaOH [53] and silk fibroin modification on the surface [54]. Besides, the hydrophobicity of PHBHHx or PHBV/PLA fibers could be utilized to avoid intensive adhesion of some tissues to other tissues. Therefore, these fibers could be used to suture hernia, and they can stay in vivo for a long time as a tissue adhesion prevention material [55]. Other functional units (e.g. PGA which accelerates degradation) can be incorporated into the material to further improve mechanical properties.

5 Conclusion

In this work, PHBV/PLA and PHBHHx strands were evaluated in terms of appearance, mechanical properties, biocompatibility and biodegradation. PHBHHx strand showed high tensile strength, elasticity and biocompatibilty. PHBV/PLA strands showed great biocompatibility as PHBV and high processability as PLA. Thus, PHBHHx and PHBV/PLA fibers both are appropriate for use as medical sutures.

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